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FOREWORD

This book contains the complete set of papers presented at the Third United States-Japan Science Cooperation Seminar on the Subject of Holographic Imaging and Information Processing, held at Hawaii, January 8 to 13, 1973, under the joint sponsorship of the National Science Foundation and the Japan Society for the Promotion of Science. The papers present the latest advances and state of the art in areas ranging from ultrasonic holography, radar and sonar to ultrasonic pulse-echo imaging, acoustic microscopy and image processing in biomedical engineering. Computer processing of ultrasonic images is extensively discussed. Several papers describe the remarkable applications of ultrasonics in medical diagnostics. One describes new features in R & D management that are particularly relevant to the field as a result of recent changes in national science and technology policies. We also present the most recent advances in the powerful methods of image improvement (sharpening, deblurring) which have been made possible with the aid of holograms used as computing elements; the holographic image deblurring method is capable of an astonishing improvement of images that are still unattainable with even the most powerful digital computer methods. The book follows the 1971 Plenum Publishing Company book, "Applications of Holography", which comprised the papers presented at the Second U. S.-Japan Seminar held in Washington, D. C., October 13 to 18, 1969. In both these seminars, the discoverer of holography, Professor Dennis Gabor who received the 1971 Nobel Prize in Physics for his work, was an active participant. All three seminars (the first having taken place in Japan from October 2 to 6, 1967) were sponsored jointly by the National Science Foundation and the Japan Society for the Promotion of Science. The singular role of Gilbert B. Devey, Program Director, National Science Foundation, Washington, D. C., in stimulating the three meetings is noted with much gratitude.

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19. This value also holds when Bragg diffraction is used.

20. S. Ishihara and K. Sakurai, Fall Meeting of Jap. Soc. of Appl. Phys. (1972) IP-B-9

21. J. Landry, R. Smith, and G. Wade, Acoustical Holography 3, (Plenum Press, 1971) 47.

PRESENT ASPECTS OF "ULTRASONOTOMOGRAPHY" FOR MEDICAL DIAGNOSTICS

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I. HISTORICAL REVIEW OF ULTRASONIC DIAGNOSTICS

Penetration Methods - The beginning of research on the use of ultrasound for medical diagnostics was not so old as that on the use of ultrasonic energy for medical purpose. In 1942,[1] Dussik in Austria reported the possibility of diagnostic use, and in 1947,[2a] he made public a sort of ultrasonic image of some brain tumors, which he named "hyperphonogram" (Figure 1). The upper picture is for normal, and the lower is for Astrocytoma, a sort of brain tumor. Although the images are obscure in comparison to current ultrasonotomograms, the historical significance is sharply dominant.[2b] Dussik obtained these images by using ultrasonic beams of 1.2 to 1.5 MHz, which penetrated the human brain and showed varied attenuation, route-by-route, to reach the receiving transducer that is placed at the corresponding opposite side of the head. The dark patterns are for good penetration, and a simple pattern in the normal brain is the ultrasonic projection of the intracranial ventricle. In 1950,[3] Ballantine and Bolt, in the United States, made public a similar experiment with knowledge of ultrasonic attenuation in biological tissues. In 1951,[4] they made the instrumentation of this penetration method and named it "ultrasonic ventriculography."

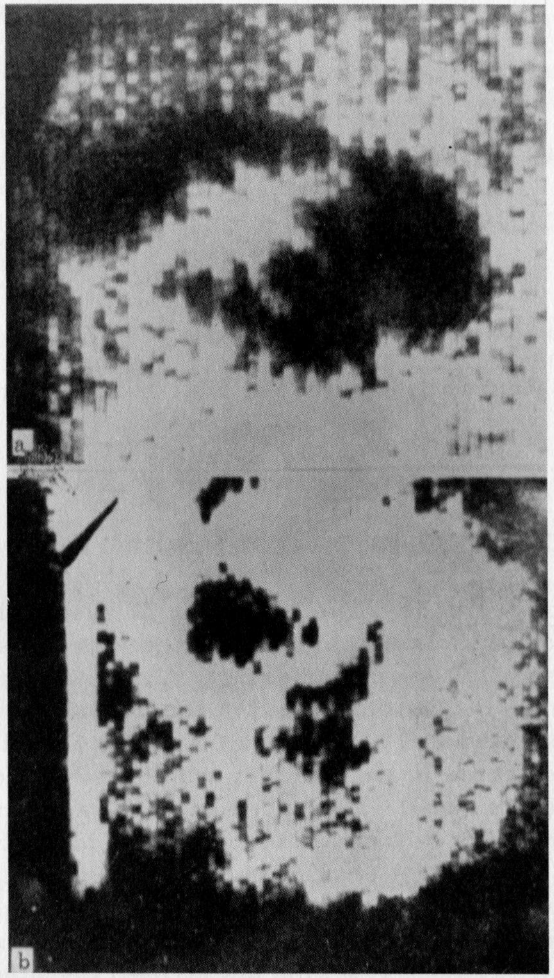

Figure 1. "Hyperphonograms" of the Brain (Dussik)

© Dussik, K. T.* : Ultraschallanwendung in der Diagnostik und Therapie der Erkrankungen des zentralen Nervensystems, Ultraschall in der Medizin, p. 283, Zurich, Hirzel, 1949. Related figure appears in the last part of the book. *Ref. Dr D. N. White, Queen's University, Kingston, Ontario, Canada; he quoted the same figure in his book entitled "Ultrasonic Encephalography," p. 2, Figure 1-1.

This method, however, has not yet been used in medical clinics because the localization of brain tumors is too indirect to be used. The main reason is that ultrasonic attenuation in tumor tissues is not so different from the attenuation in normal brain tissues, so the tumor boundaries are not clearly shown on the hyperphonograms. In addition, ultrasonic attenuation at the skull varies place-by-place so much that the variation is generally dominant over the useful attenuation difference between the pathological and normal tissues. This fact results in masking of the objective pattern. Moreover, as Güttner[5] in Germany pointed out, the large attenuation in the skull, as large as 40 to 50 dB/cm, might be an additional barrier in the ultrasonic technique relevant to the method.

Echo Methods; A-Scope - In 1950, French and Wild[6] in the United States, tried an ultrasonic pulse-echo method and succeeded in diagnosing brain tumors. Figure 2 depicts the first A-scope representation of ultrasonic echoes from the inside of a brain tumor.[6] The echo pattern is quite different from that of normal tissues. In 1951, Kikuchi, et al.[7] reported the ultrasonic detection of hematoma in the brain. They used equipment for ultrasonic flaw detection which had been developed for industrial use at that time. Through some improvements, they were soon successful in detecting various brain tumors through the skull. As it had been said that ultrasonic waves in general have biological effects on living tissue, they made various animal tests to determine if the ultrasonic pulse used for a diagnostic purpose would affect the animal's brain.[8] After physiological confirmation that there were no effects, the application of ultrasonic examination has become clinical routine for the first time in Japan. Through hundreds of further investigations and development, clinical methods have been established.[9]

In the United States, however, the U. S. Atomic Energy Commission[10] made public a report in which it was stated that ultrasonic reflection methods as well as penetration methods were unsuitable for detection of brain diseases.

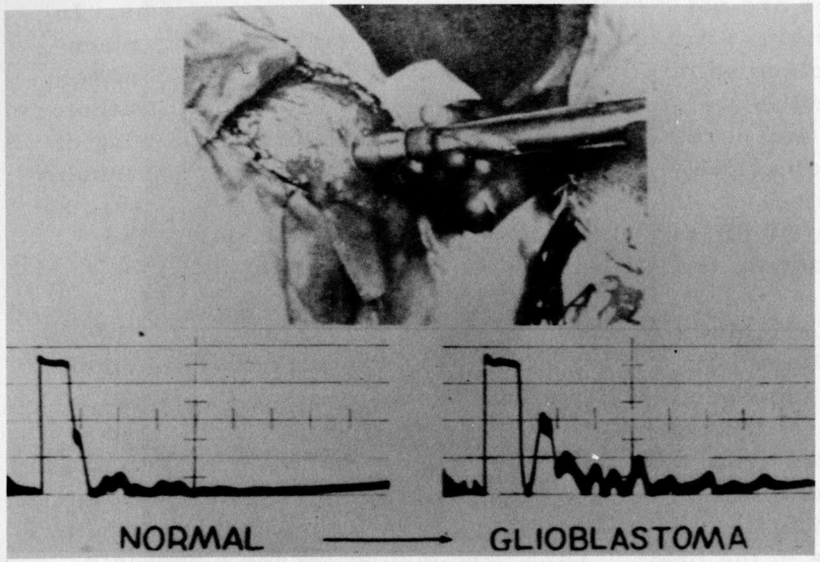

Figure 2. A-mode Echo Patterns Obtained for the First Time by French and Wild.

© Wild, J.J.* and Reid, J.: The effects of biological tissues on 15 mc pulsed ultrasound, J.A.S.A., Vol.25, No.2, March, 1953, p.270, p.276, Figure 7.
* J.J. Wild, Department of Electrical Engineering, University of Minnesota, Minneapolis, Minnesota, USA.

It might not be denied that the Commission's report would not have delayed the general development of ultrasonic diagnostics in the United States, although the skull is really a big barrier for ultrasonic penetration. White in Canada[11] also stated the difficulty through such a statement, "If it were possible to make clinically useful tomograms (of the brain) as easily as they could be made of the pelvic organs, neurological investigation would be enormously helped and simplified." Though we[12] are at present able to obtain clinical tomograms of the brain in various tumor cases by means of a contact sector scanning from the outside of the skull, the tomograms cannot be so clear as those of other organs. With regard to brain mapping, Fry, et al.[13] reported in 1965 that ultrasonic echoes were obtained from a part of living tissue just after the part was otherwise irradiated by stronger ultrasound, and suggested that the echo sources were the acoustic impedance change of the local part due to a slight temperature difference induced by ultrasonic absorption.

In the field of cardiology, Edler and Hertz[14,15] in Lund University, Sweden, inaugurated the "Ultrasonic Cardiography." This is a sort of time-position-indication of ultrasonic echoes which come back from the inside of the pulsating heart.

The clinical application to other human organs was developed in a swift step dramatically after the pulse technique was employed in medical examinations, not only by A-mode representation, but also by B-mode, PPI, and other tomographical representations.

<u>B-Scope; Tomography</u> - Physicians in clinics were generally requiring a more instructive representation of the echo sources in the organs than that of the A-scope representation. Plan-position-indication type of representation was accordingly considered among the research workers in the field. In 1952, Wild[16] developed for the first time, the two-dimensional visualization of a section of the human body.

Howry, in Denver, Colorado, introduced an ingenious technique of a compound scanning into Wild's simple scanning and obtained clear tomograms during 1954 to 1957.[17] He named the technique "Somascope". Figure 3 shows an example of the tomograms, of a horizontal section of the neck of a patient. Scanning was as follows:

> The neck was immersed in a water vessel where an ultrasonic transducer was made traveling around the neck while the transducer was quickly swinging its directional beam.

The idea of this compound beam motion is provided to ultrasonically illuminate dark target planes which usually exist in the human tissues in relation to the incidental and reflected ultrasonic waves when the waves are to be managed by a single transducer.[18] Howry later improved the apparatus by introducing a converging beam transducer and some appropriate functions in the electronic circuits so that clearer tomograms were obtained.[17]

Another contemporary approach was going on in our laboratories where we were engaged in obtaining apparatus for clinical use. Figure 4 shows our first tomographic scanner, moving horizontally in a water vessel that is attached to the breast. Ash shown in Figure 5, a section of the breast is displayed on a cathode ray screen when the transducer makes one linear trip slowly while the ultrasonic pulses of a high repetition rate are making the echo-sounding through the mammary structure. By taking the picture of the screen with a slow exposure, or when the cathode ray screen is a sort of memory scope type, the section of the living organ is visualized as if observers were looking at the slice of the part. Such slice representation of human organs had been called "tomograms" in the X-ray examination; thus, this technique was named "ultrasonotomography".[19, 20] Figure 6 shows a pair of examples of the actual tomograms for a normal breast and a malignant one. The cancer tissue is indicated by the gathered mass of bright spots in accordance with acoustically more inhomogeneous and reflective structure of the malignant tissue

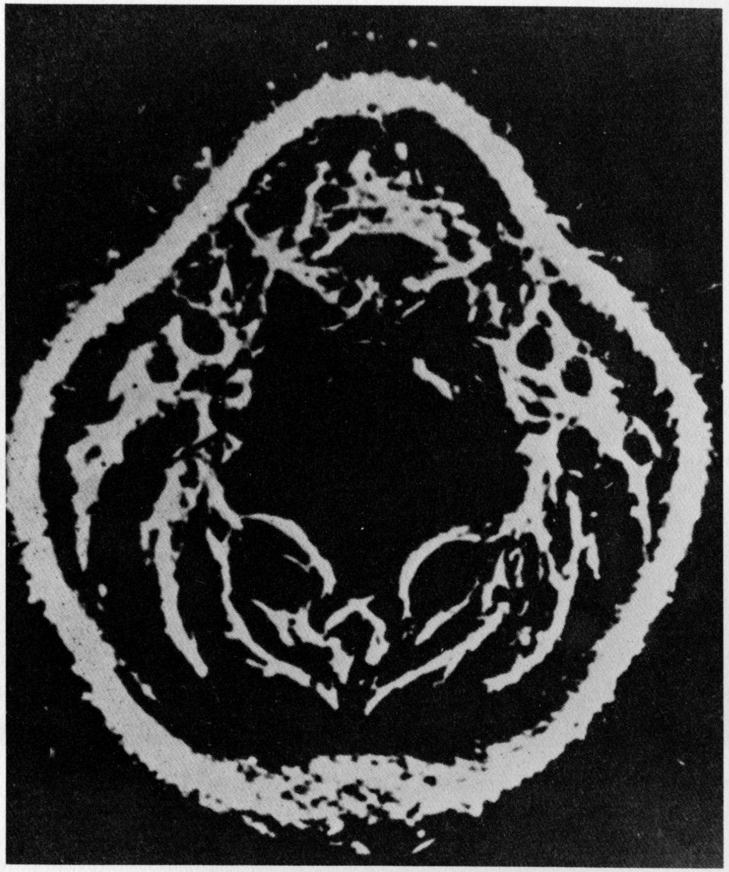

Figure 3. A Tomogram of the Neck (Howry)

© Howry, D. H. *: Techniques used in ultrasonic visualization of soft tissues; Ultrasound in Biology and Medicine (Ed., E. Kelly), p. 49, American Institute of Biological Science, Washington, D. C., 1957. P. 62, Figure 10.
* D. H. Howry, Ultrasonic Research Unit, University of Colorado School of Medicine, Denver, Colorado.

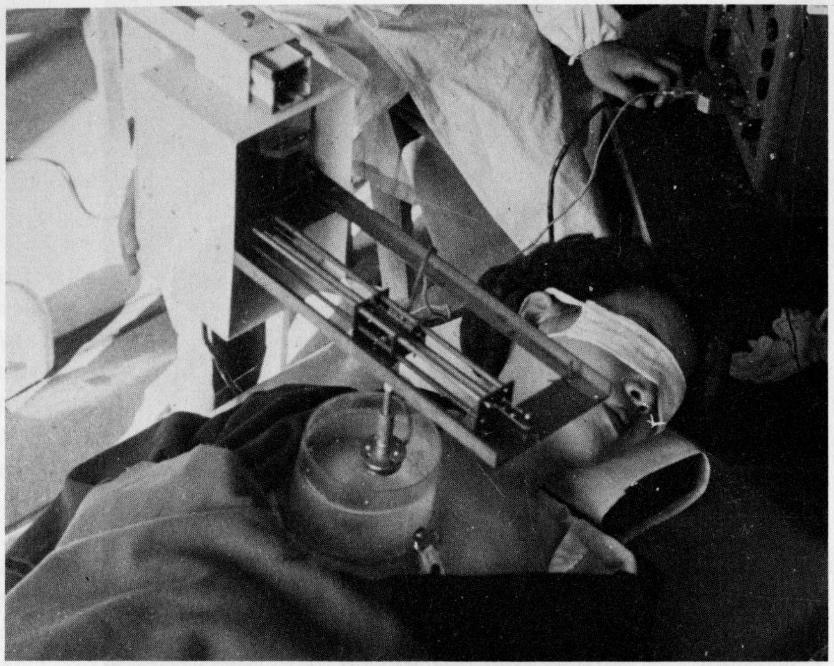

Figure 4. First Clinical Experiment of Ultrasonotomography for Breast Tumors. (Kikuchi, et al.)

© Kikuchi, Y., * Tanaka, K., Wagai, T. and Uchida, R.: Early cancer diagnosis through ultrasonics, Journal of Acoustical Society of America, 29(7), P. 824, 1957, p. 831, Figure 16. * Professor of Research Institute of Electrical Communication, Tohoku University, Sendai, Japan.

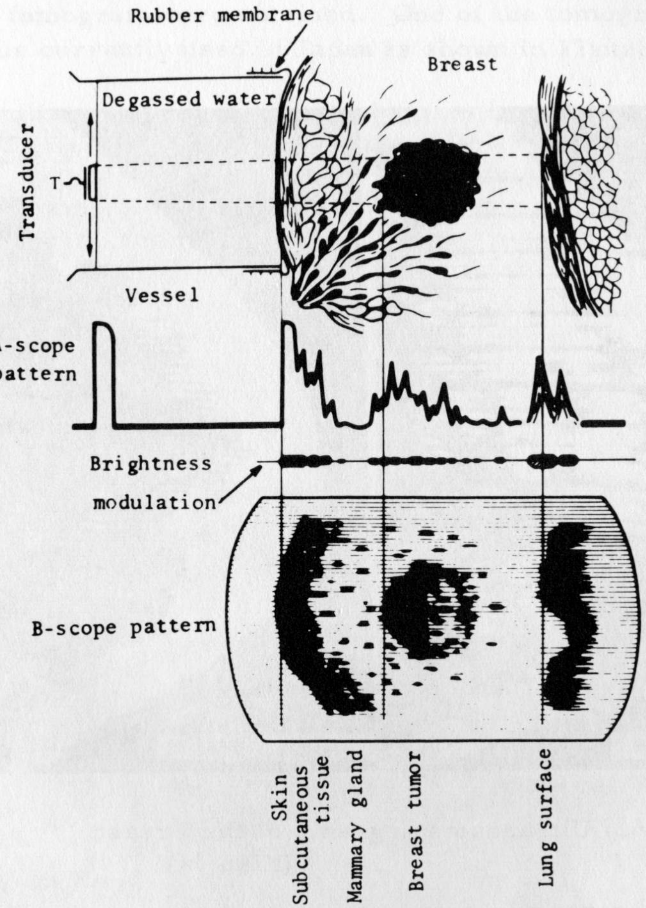

Figure 5. Illustration for Ultrasonotomography.

© Yamakawa, K.*, Wagai, T.*, Kikuchi, Y. and Uchida, R.: Application of Ultrasound in Diagnostic Fields (Tentative translation from Japanese), Inst. of Elect. Comm. Eng. Japan, Report of Professional Group in Electrical Apparatus for Medicine, Feb. 12, 1959, p. 14, figure 11.

* Juntendo U. School Med., 1-1, 2-Chome, Hongo, Bunkyo-Ku, Tokyo, Japan.

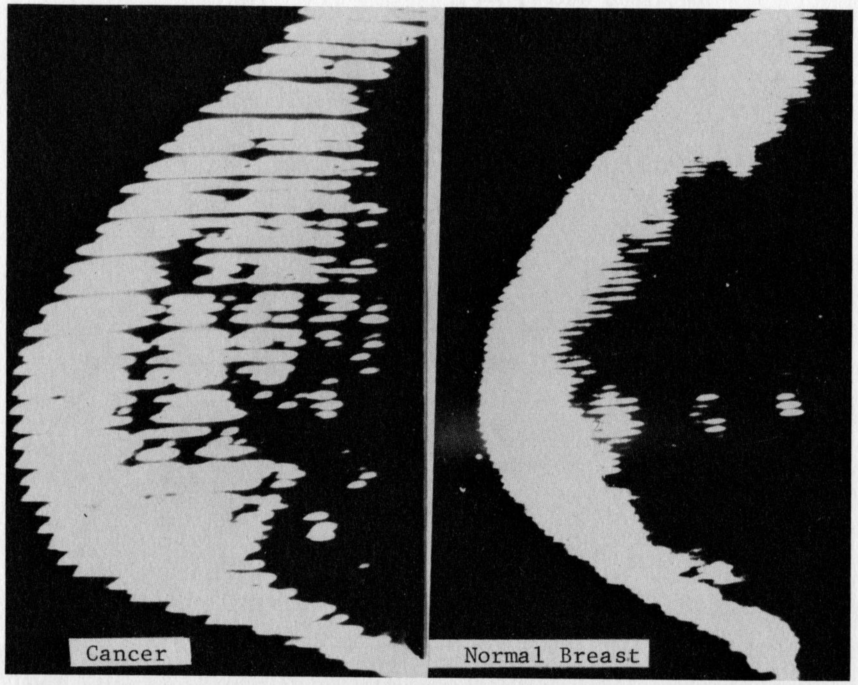

Figure 6. Ultrasonotomograms of the Breast. (Kikuchi)

Improved apparatus are widely used at present by which clearer tomograms are obtained. One of the tomographic apparatus currently used in Japan is shown in Figure 7.

Such ultrasonotomographic techniques have been extended to various fields of clinical medicine. In 1962, Kossoff and others[21] reported its application to obstetrics, and showed the tomograms of the fetus living in the uterus. In Figure 8 an example is shown of the tomograms presently obtainable

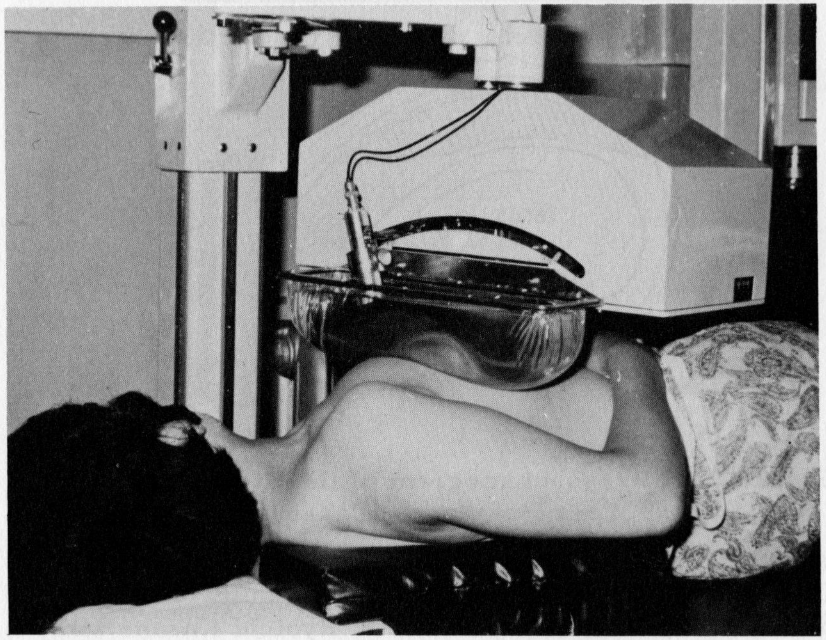

Figure 7. A Tomographic Apparatus Currently used in Japan.

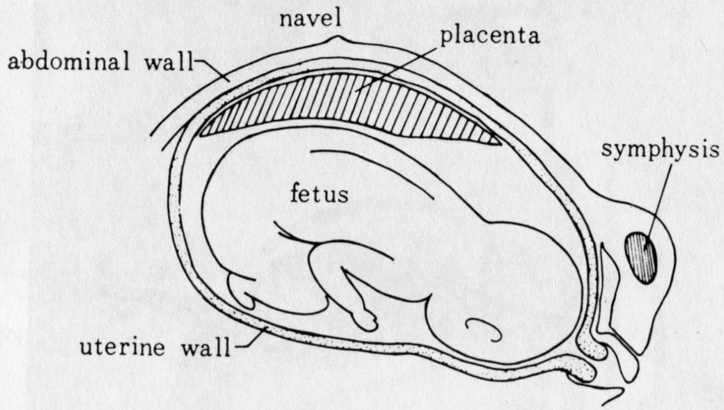

Figure 8. Ultrasonotomogram of the Uterus with 38 Week Pregnancy (Takeuchi)

© Kusano, R. *, and Takeuchi, H. *: Application to Obstetrical Field, Medical Apparatus Culture, 11, 64 (1970-8), P. 65, Figure 8. * Ryoichi Kusano and Hisaya Takeuchi, Juntendo University School of Medicine, 1-1, 2-Chome, Hongo, Bunkyo-Ku, Tokyo, Japan.

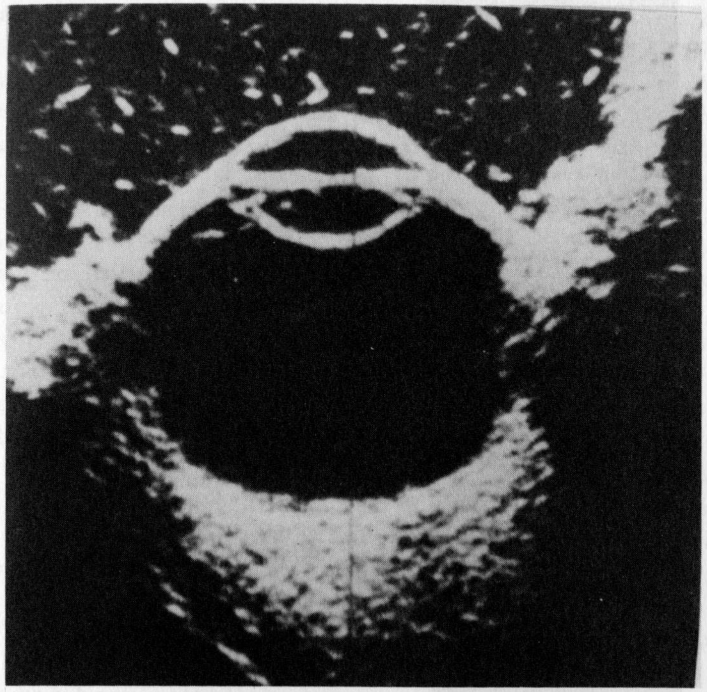

Figure 9. Ultrasonotomogram of the Human Eye (Baum)

© Baum, G. *: A Comparison of the merits of scanned intensity modulated ultrasonography versus unscanned A-scope ultrasonography, Diagnostic Ultrasound (Ed: C. C. Grossman), P. 59, Plenum Press, New York, 1966, P. 62, Figure 1. * Gilbert Baum, M. D., Albert Einstein College of Medicine, Yeshiva University, Bronx, New York, USA.

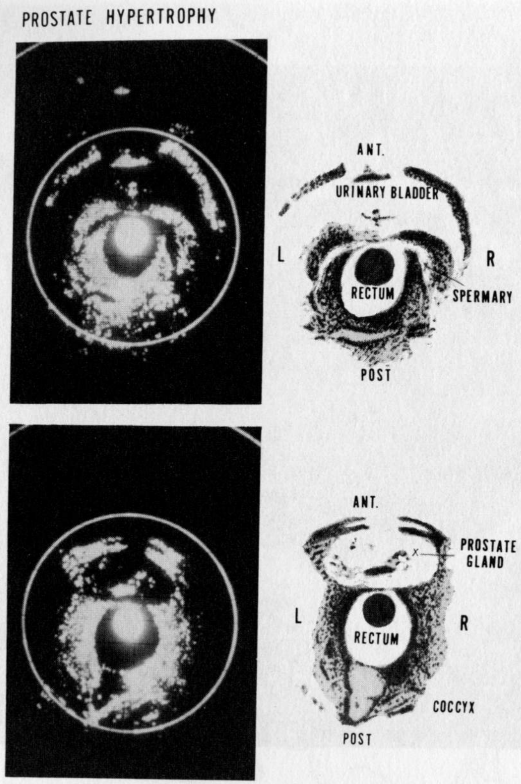

Figure 10. PPI Tomograms of the Urinary Bladder and Prostate Gland (Tanaka, et al.)

© Tanaka, M.*, Oka, S., Watanabe, H., Kato, T., Shishido, S., Ebina, T., Kosaka, S., Terasawa, Y., Unno, K., Nitta, K., Kikuchi, Y., Okuyama, D., Uchida, R.: Ultrasono-tomography of the intrapelvic organs(a tentative title translated from the Japanese), Proc. 13th Meeting Japan Society of Ultrasonic Med., 79, 1968, p. 80, Figure 2.

* Motonao Tanaka, M. D., Res. Inst. Tuberculosis, Leprosy and Cancer, Tohoku U., Sendai, Japan.

in any clinic. In 1958 through 1965, Baum[22,23] was engaged in the ultrasonic diagnostics in the field of ophthalmology, and obtained tomograms of the eye as shown in Figure 9, as a typical example. The cornea, iris, retina, etc., are clearly visualized in a living human eye. In 1957, Wild and Reid[16] reported an equipment for obtaining three-dimensional information from the lower bowel. The transducer revolves inside the rectum to obtain plan-position-indication of a section. The sectional plane is then shifted to the next successively. This technique is useful at the present time for the examination of the urinary bladder and prostate gland.[24,25] In Figure 10, examples are shown, which were obtained in our clinic, of a case of prostate hypertrophy with stones. By a similar technique, we have developed intra-esophageal[26] and intratracheal[27] methods of ultrasonic examination. The height of ultrasonic frequency for these methods of examining tubular organs should be 10 to 15 MHz, as the transducers must be smaller in size; and in addition, we recommend concave transducers so as to use a narrower and nearly focusing ultrasonic beam. We have already proposed a suitable design procedure for the concave transducer. The outline will be described later in this paper.

Through many contributions given by research workers in the world, ultrasonic examination for medical diagnostics has been established in almost all fields of clinical medicine; the heart is the only organ that did not allow anyone to take its ultrasonotomogram because every part of a living heart is always pulsating with faster velocity than to be assumed stationary in any case of usual scanning.

Kikuchi and others[28] have been engaged in this problem since 1966, and introduced a technique named "ultrasono-cardio-tomography" which is based on a method of display in synchronization of the heart movement. They have further developed "Kineto-Ultrasonotomography"[29] which is a sort of animation of the individual cardiotomograms. The details of these methods are described in a separate paper.

II. ACOUSTICAL PROPERTIES OF BIOLOGICAL TISSUES

In any case of ultrasonic visualization of biological tissues, the acoustical properties of the tissues such as the propagation velocity and attenuation of ultrasonic waves in the tissues and the acoustical impedance (ρc) of the tissues are important factors. Especially, in the pulse-echo method, the acoustical impedance difference among the relevant tissues determines the echo intensities. As to the propagation attenuation, not only does it have close relation with the echo intensities, but also is one of the important factors in the medical diagnostics for brain edema, hematoma, cancer, epilepsy of certain cases, breast tumors, etc.[8a, 30-33]

Many research workers have been engaged in the measurement of the acoustic properties of biological tissues. In the following, some summarized data are shown: Table 1 shows the data obtained by Kikuchi and others[7b, 32] in 1952, giving sound velocity, c, specific gravity, ρ, acoustic impedance, ρc, of various tissues, and also the estimated percentage reflection of a plane wave when it is incident normally to the interface between any two tissues. Note that the reflection is usually very small.

With regard to the ultrasonic absorption, there are usually appreciable differences between the normal and tumor tissues. Figure 11 is an example of those in the brain.[34] Wells[35] has published a similar table of interface reflection, including more biological materials. There are, however, no data concerning pathological tissues. In 1956, Goldman and Heuter[36] reported a condensed data available at that time on the velocity and absorption of high frequency sound in mammalian tissues. In Tables 2 and 3, some essence relevant to human bodies is shown; in Table 2, the sound velocities of various human tissues; and in Table 3, the sound absorptions in the blood, plasma and in normal and various tumor tissues. The authors said in the report that the absorption coefficient of most tissues lies in the range from 0.5 to two dB/cm/MHz, and with other data which are not shown here, the authors showed an estimated frequency dependence of it as shown in Figure 12.

Table 1

Acoustical properties of human tissues

Tissue	Sound Velocity (c) cm/s	Density (ρ) gr/cc	Acoustic Impedance (ρc) ab·ohm	Power Reflection Coefficient in % (Normal incidences of plane waves)							
				air	steel	water	fresh cerebrum (cow)	glioma	meningioma	cerebellum	cerebrum
	$\times 10^5$		$\times 10^6$								
cerebrum	1.53	1.038	0.159	99.9	86.9	0.27	0.00001	0.046	0.0023	0.048	0
cerebellum	1.47	1.034	0.152	99.9	87.4	0.093	0.049	0.00001	0.029	0	
meningioma	1.49	1.056	0.157	99.9	87.0	0.23	0.0026	0.028	0		
glioma	1.46	1.042	0.152	99.9	87.4	0.095	0.048	0			
fresh cerebrum (cow)	1.54	1.032	0.159	99.9	86.9	0.28	0				
water	1.43	1.00	0.143	99.9	88.1	0					
steel	5.88	7.7	4.53	100	0						
air	0.33	0.0012	0.00004	0							

Power Reflection Coefficient: $R_o = \left(\frac{\rho_1 c_1 - \rho_2 c_2}{\rho_1 c_1 + \rho_2 c_2}\right)^2$

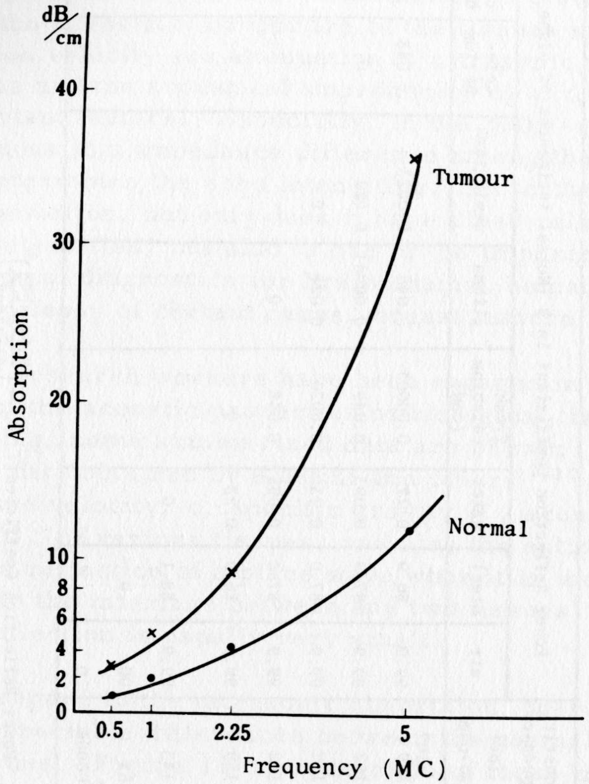

Figure 11. Sound Absorption in Brain Tissues
Upper: Brain Tumour Tissue
Lower: Normal Brain Tissue

© Tanaka, K., * Kikuchi, Y. and Uchida, R.: Ultrasonic Detection of anatomical abnormalities in the human cranium, Part 2. Reports of 1953 Spring Meeting, Acoustical Society of Japan, No. 2-9, May, 1953 (in Japanese) Page 54, Figure 3. * Kenji Tanaka, Professor Emeritus of Juntendo University School of Medicine, 2-1-1, Hongo, Bunkyo-Ku, Tokyo, Japan.

Table 2
Velocity of H. F. Sound in Human Tissues

Tissue	Temp. °C	Frequency MHz	Velocity m/s
muscle	24	1.8	1568
liver	24	1.8	1570
fat	24	1.8	1476
limb	body	2.5	1540
meningioma	body	2.26	1540
skull bone	body	0.8	3360
breast carcinoma	24	1.8	1573

© Goldman, D. E. * and Hueter, T. F. ** Tabular data of the velocity and absorption of high-frequency sound in mammalian tissues, Journal Acoustical Society America, Volume 28, No. 1, P. 35, 1956. P. 35, Table I.
* D. E. Goldman, Naval Medical Research Institute, Bethesda, Maryland.
** T. F. Hueter, Massachusetts Institute of Technology, Cambridge, Massachusetts.

Table 3

Abosrption of h. f. sound in human tissues

Tissue	Condition	Frequency MHz	Absorption* cm^{-1}
plasma	refrig.	1.7	0.04
plasma	---	1.0	0.007
blood	---	1.0	0.02
muscle	---	0.80	0.1
brain	fixed	1.7	0.18
brain	fixed	3.4	0.37
medulla oblongata	---	1.7	0.14
medulla oblongata	---	3.4	0.34
liver	20 hr post mortem	1.	0.15
liver	do	3.	0.23
liver	do	5.	0.35
fat	melted	0.87	0.045
fat	melted	1.7	0.09
fat	melted	3.4	0.16
skull bone	fresh or fixed	0.6	4.5
skull bone	---	1.2	17.
skull bone	---	2.25	53.
skull bone	---	3.5	80.
sciatic nerve	----	3.4	0.35

* Absorption coefficient α in $A = A_o e^{-\alpha x}$ where A_o is an amplitude of sound.

© Goldman, D.E.* and Hueter, T.F.** : Tabular data of the velocity and absorption of high-frequency sound in mammalian tissues, Jour. Acoust. Soc. Am., Vol.28, No.1, p.35, 1956. p.36, Tab.II.

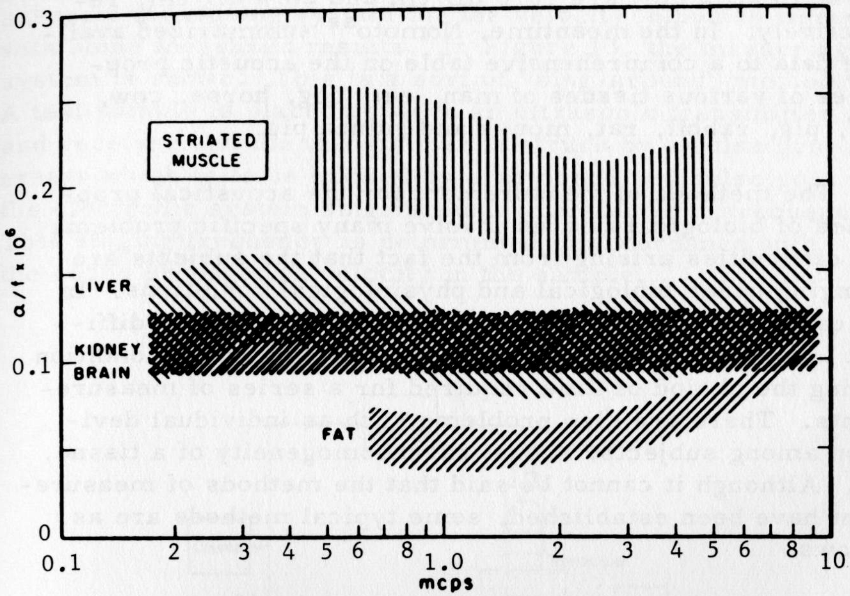

Figure 12. Estimated Relation Between Sound Absorption and Frequency for Several Mammalian Tissues. (Goldman and Hueter)

© Goldman, D. E. * and Hueter, T. F. ** Tabular data of the velocity and absorption of high-frequency sound in mammalian tissues, Journal Acoustical Society America, Volume 28, No. 1, Page 35, 1956. P. 37, Figure 1.

* D. E. Goldman, Naval Medical Res. Inst., Bethesda, Md.

** T. F. Hueter, Massachusetts Institute of Technology.

In 1964, Ishikawa[37] reported that the absorption in the brain tissues is different between the sagittal direction and the horizontal direction toward the forehead. At five Megahertz they are 7.2 dB/cm and 10.7 dB/cm, respectively; and at 10 MHz they are 16.6 dB/cm and 28.4 dB/cm, respectively. In the meantime, Nomoto[38] summarized available data to a comprehensive table on the acoustic properties of various tissues of man, cat, dog, horse, cow, calf, pig, rabbit, rat, mouse and guinea pig.

The methods of measurement for the acoustical properties of biological tissues involve many specific problems and difficulties arising from the fact that the subjects are changing under biological and physiological conditions. In the case of sample measurement, it is considerably difficult to keep the samples under a certain biological condition during the period of time required for a series of measurements. There are also problems such as individual deviation among subjects, acoustical inhomogeneity of a tissue, etc. Although it cannot be said that the methods of measurement have been established, some typical methods are as follows:

After Pohlman,[39] Hueter and others,[40] in 1948, placed an excised biological sample in a cylindrical glass tube filled with water and measured acoustical absorption of the biological tissue from the change of acoustical power reaching the other end of the tube. In later years (1952 and 1953), Kikuchi and others[34] used pulsed ultrasonic waves. Yoshioka and others[41] recently have made an instrumentation of the method with specific care. Ishikawa[37] used data on the temperature rise of tissues for the observation of ultrasonic absorption. The temperature rise with regard to the lapse of time is a suitable measure to indicate the relative absorption of biological tissues. Carstensen[42] has proposed a suitable method to measure liquid absorption.

With regard to the measurement of ultrasonic velocity in biological tissues, most research workers[7,43,44,32] employed some sort of ultrasonic interferometer and/or

method of multiple reflection principle, placing a tissue or a biological liquid sample between two parallel planes, one of which or both of which were transducer planes. An ultrasonic thickness gauge was also used as a variation of interferometer principle. Venrooij[45] recently reported an accurate measuring system for the velocity in human tissues with some measured results. In Figure 13, the measuring system is shown. This is a sort of "sing-around" method.[46] A test sample is placed between an ultrasonic transmitter and receiver and the transmitter is struck by a pulse generator which is to be excited by every received pulse so the electronic system sings around with a definite frequency. This singing frequency is determined in accordance with the sound propagation velocity in the sample.

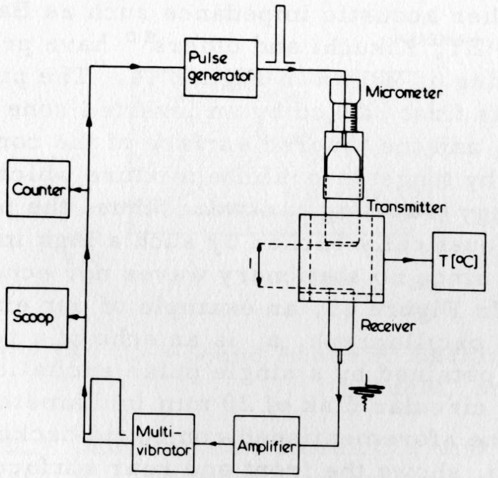

Figure 13. Velocity Measuring System Along the Sing-Around Principle (Venrooij)

© Van Venrooij, G.E.P.M.*: Measurement of ultrasound velocity in human tissue, Ultrasonics, Page 240, October, 1971. P.240, Figure 1.

* Van Venrooij, G.E.P.M., Department of Neurosurgery, Medical Physics Department, State University, Utrecht, The Netherlands.

III. RECENT RESEARCH RESULTS ON THE GENERATION AND RECEPTION OF PULSED ULTRASOUND

1. Distance Resolution - In general pulse echod methods, distance resolution depends directly on the length of the wave train; that is, the pulse length. There were many reports in the past on this problem, and convenient circuits[47,48] for the generation of pulsed electrical oscillation have been suggested. On the side of ultrasonic transducers, a means in which a piezoelectric plate is backed by a high damping material has been recommended.[49] For the backing material, a mixture of artificial resin and fine metal powder such as tungsten has been used. Available highest acoustic impedance is around 10×10^6 $Kg/m^2.s$, and is nearly sufficient for damping the free oscillation of a quartz and/or LH transducer. For other transducers made of piezo-active material of higher acoustic impedance such as $BaTiO_3$-ceramic, and/or PZT, Kikuchi and others[50] have proposed a compound backing as shown in Figure 14. The piezo-active ceramic plate is first backed by an inverted cone or wedge made of brass, and the tapered surface of the cone or wedge is then backed by tungsten-araldite mixture which absorbs ultrasonic energy traveling backward; thus, the piezo-active plate can be acoustically backed by such a high impedance as brass itself since no stationary waves nor echoes yield in the brass. In Figure 15, an example of our experiment is shown. The oscillograph, a, is an echo of a reflective plane in water obtained by a single pulse excitation given on a 2.25 MHz circular disk of 20 mm in diameter, made of PZT, with the aforementioned compound backing. The oscillograph, b, shows the front and rear surface echoes of a rubber plate, the thickness of which is as thin as 1.5mm.

In recent years we[51] also proposed an active damping which is based on a doubled electrical excitation. A piezo-active disk is struck by two electrical pulses, one of which is delayed by an electrical network so the natural mechanical ringing of the disk is forced to diminish immediately after the first wave.

Recently, we[52] have established another method of radiation and reception of very short ultrasonic pulses by using a quarter wave-length layer. This idea was first proposed by McSkimmin[53] and developed by Kossoff[54] to some extent. We[55,56] then developed an engineering procedure to determine the optimum impedance of the quarter wave length layer material. The procedure enables us to easily calculate the train of ultrasonic pulses, the resultant pulse-train width and the sound intensity in relation to the layer materials. A material of any desired acoustic impedance is not easily available among markets so we recommend use of a suitable mixture of artificial resin and metal powder. Though the mixture is originally absorbing material, the propagation loss in a thin layer, as thin as a quarter wavelength, is usually very small. In any choice of material,

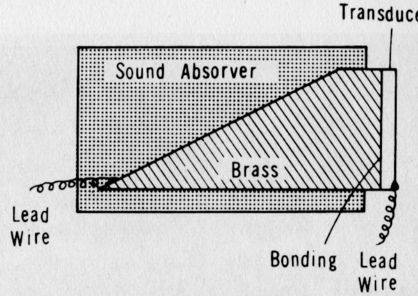

Figure 14. Compound Acoustic Backing (Kikuchi, et al.)

* Y. Kikuchi, Professor of Research Institute of Electrical Communication, Tohoku University, Sendai, Japan.

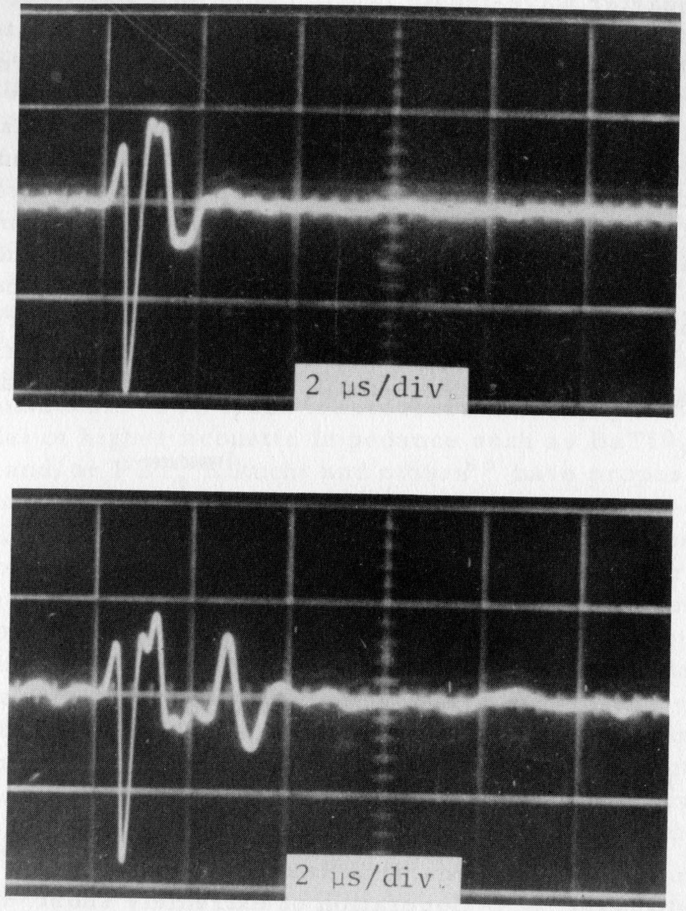

Figure 15. Echoes of High Resolution

© Kikuchi, Y., Okuyama, D., Kasai, C. and Tatebayashi, K.: Generation of extremely short ultrasonic pulses into liquid by acoustic backing transducer. Reports of Japan Society of Ultrasonics in Medicine, 17th Meeting, P. 53, 1970 (in Japanese). P. 54, Figures 2c and 3b.

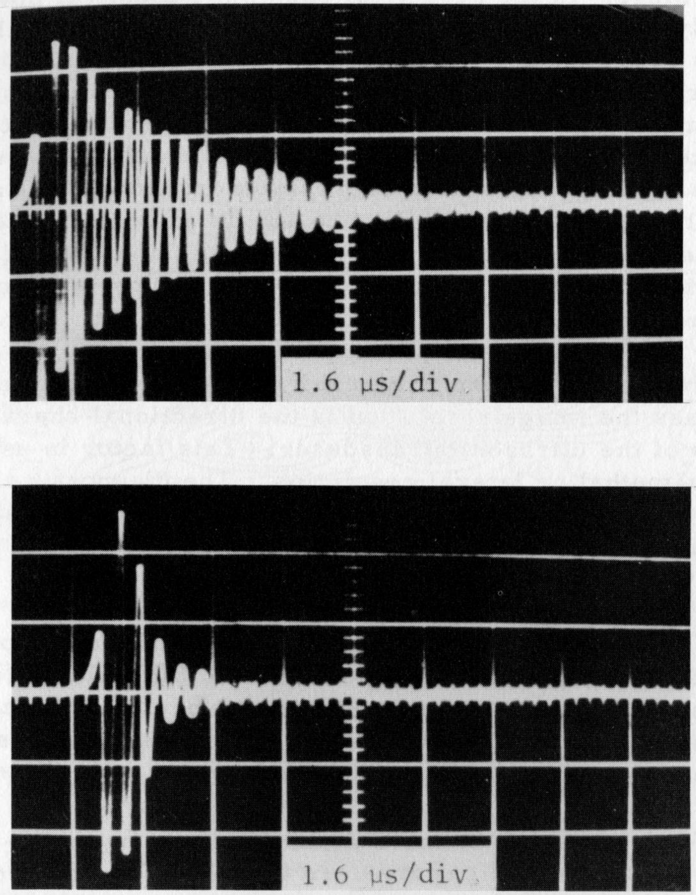

Figure 16. Effect of a Quarter-Wavelength Layer

© Kikuchi, Y., Okuyama, D., and Kasai, C.: Generation of short ultrasonic pulses by means of ultrasonic transducer with an intermediate layer of a quarter wavelength. Reports of Professional Group on Ultrasonics, Institute of Electronics Communications Engineers, Japan, US-71-13, 1971 (in Japanese). P. 6, Figure 9.

the sensitivity of the transducer as a whole is much higher than that of the aforementioned transducers with the acoustical backings. In the acoustical backing, acoustical power is largely absorbed, and only a small portion of the total power is sent forward, to be used as acoustic waves. On the contrary, the quarter wavelength device does not dissipate acoustic power but does transfer the power completely to the acoustic medium, or does receive the acoustic power from the medium through an appropriate transforming action of acoustic impedances. In Figure 16, an example of the experimental results is shown. The ringing of a PZT disk such as is shown in Oscillograph a can be damped as shown in b with a favorable increase of amplitude.

2. Azimuthal Resolution. - Another important factor that determines the image resolution is the directional characteristics of the ultrasonic transducer. This factor is usually called azimuthal or lateral resolution. The diagnostic application of ultrasound is generally confined within the near field zone of the transducer. This zone is often called Fresnel's interference zone, and the axial length of the zone for a circular disk transducer[57,58] of Radius R is about R^2/λ, where λ is the wavelength of the radiated sound. So far as pulse-echo method is concerned, the field pattern of ultrasonic waves within this region can be usually approximated as a beam of 2R in diameter, with parallel waves of constant intensity.

With regard to concave disks for focusing transducers, the near-field patterns are theoretically derived by O'Neil,[59] Torikai[60] and others. In Figure 17, an example is shown in which a dimensional parameter, D, is chosen as 10. And $D = R^2/a\lambda$, where a is the focal length of the concave disk. In Figure 18, the sound field along the beam axis is shown with D as parameters. The horizontal axis Z is the distance along the sound beam which is normalized by the focal length, a. Kikuchi and others utilized Torikai's result and have recommended that the dimensional parameter D should be around four, so as to obtain optimum transducers for diagnostic use.[61] In this case, "optimum" means that the lateral three dB width of

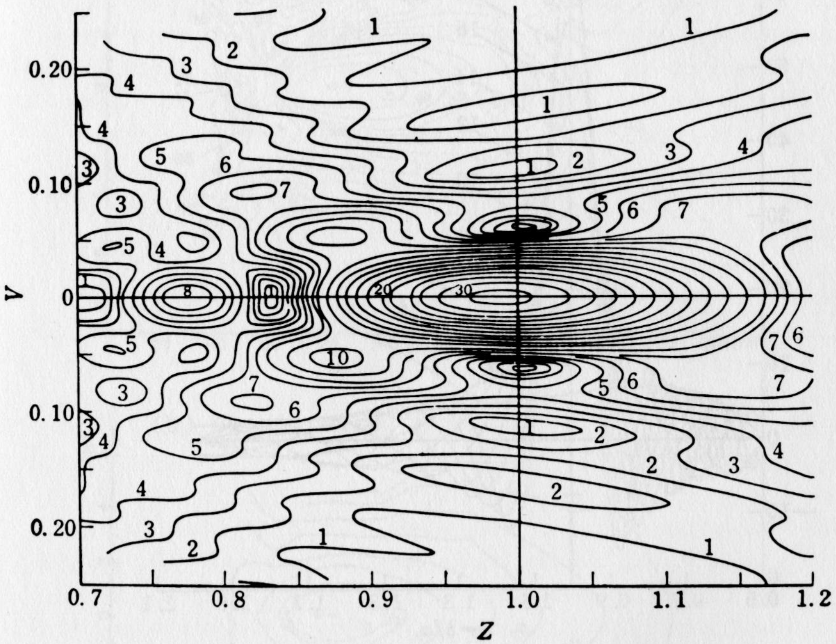

Figure 17. Near-field Pattern of a Focusing Transducer when $D = R^2/a\lambda = 10$ (Torikai)

© Torikai, Y.*: Cho-onpa Gijutsu Binran (A Handbook of Ultrasonic Engineering). Edited by J. Saneyoshi, Y. Kikuchi and O. Nomoto. Page 1418, Fig. 1-27, Nikkan Kogyo Press Ltd,** Tokyo, Japan.

* Yasuo Torikai, Professor of Institute of Industrial Science, University of Tokyo, Roppongi, Tokyo.

** Nikkan Kogyo Press, Ltd, 8-10, Kita 1-Chome, Kudan, Chiyoda-Ku, Tokyo, Japan.

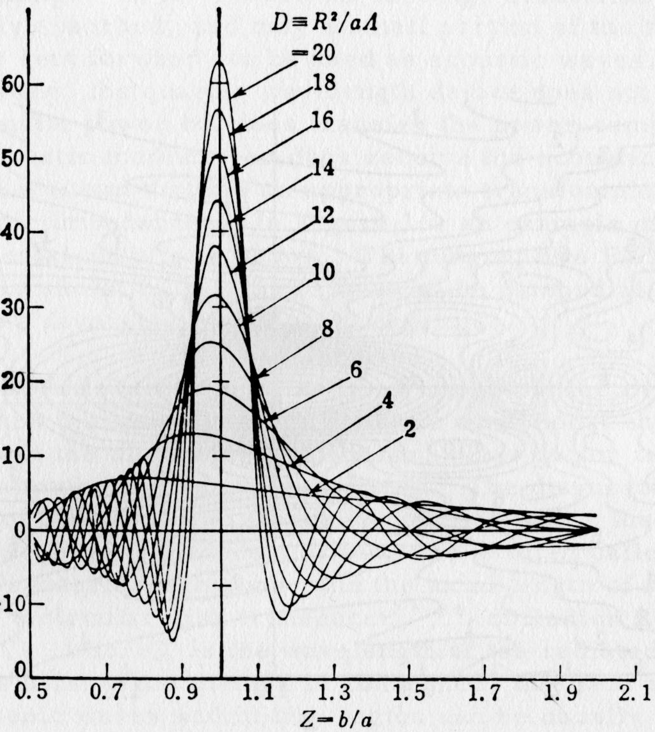

Figure 18. Sound Field on the Beam Axis of Concave Disks, $D = R^2/a\lambda$ being Dimensional Parameters. (Torikai)

Ⓒ Torikai, Y.: Calculations of near sound field (tentative translation from Japanese), The Institute of Electronics and Communication Engineers of Japan. Report of Professional Group in Ultrasonics, November, 1962, PP 17-18, Figure 7.

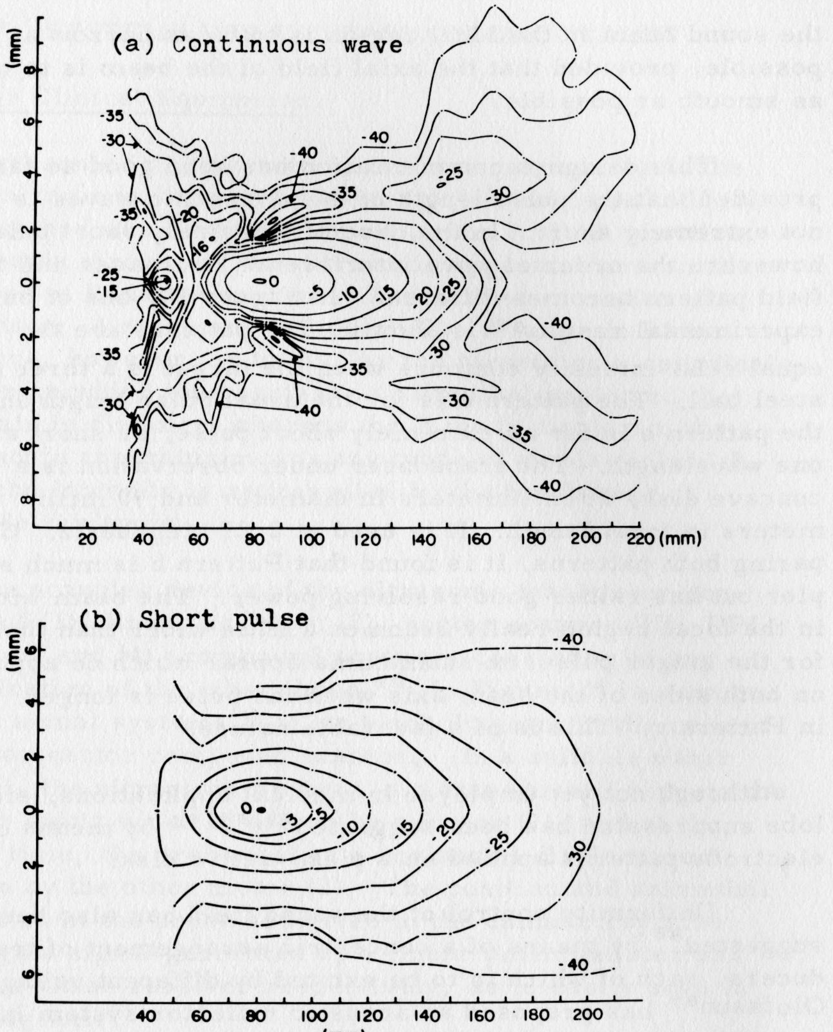

Figure 19. Near-field Patterns of a Concave Disk (a) Continuous Waves; (b) Extremely Short Pulse. (Kikuchi, et al.)

Ⓒ Kikuchi, Y., Okuyama, D. and Kasai, C.: Sound field characteristics of extremely short ultrasonic pulse radiated by a concave transducer. Reports of Japan Soc. Ultrasonics in Med., 19th Meeting, 19-31, p. 61, 1971 (in Japanese). P. 62, Figure 2.

the sound beam in the focal region is to be as narrow as possible, provided that the axial field of the beam is to be as smooth as possible.

This design recommendation has been good so far, provided that the pulse length of the ultrasonic waves is not extremely short. In the case of extremely short pulse, however, the order of wave interference decreases and the field pattern becomes different. In Figure 19, one of our experimental results[62] is shown. The patterns are the equal echo-intensity contours when the target is a three mm steel ball. The pattern a is for the usual pulse length and the pattern b is for an extremely short pulse, as short as one wavelength. The transducer under observation is a concave disk, 25 millimeters in diameter and 90 millimeters in focal length. It is used at 2.25 Megahertz. Comparing both patterns, it is found that Pattern b is much simpler but has rather good resolving power. The beam width in the focal region really becomes a little wider than that for the longer pulse; no submaxima appear which do appear on both sides of the beam axis when the pulse is longer, as in Pattern a. This is of a favorable nature.

Although not yet employed in medical applications, side lobe suppression has been suggested[63,64,65] by means of electrode patterns applied on a piezo-active disk.

Uniformity control of the sound field has also been suggested[66] by means of a concentric arrangement of transducers, each of which is to be excited by different voltage. Olofsson[67] has proposed an acoustic reflector system in place of concave disks of large size. The azimuthal resolving power in the focal region is extraordinarily excellent, but the axial field distribution is too rough to be used in the usual way. Fry and others[68] then used this mirror system through a computer control in the three-dimensional sector-scanning mode. An idea of composite echograms was suggested with a time-gated display and other relevant techniques.

IV. SYSTEMS FOR ULTRASONOTOMOGRAPHY

1. For Clinical Equipment:

Ultrasonic imaging used in medical diagnostics at the present time is based mainly on a pulse-echo method. As this method was originally developed in RADAR practice, electronic circuits of the ultrasonic apparatus are similar to those of Radar equipment. There are two modes of operation: A mode and B mode. The B scope in Radar practice, however, is to display the targets on a coordinate system in which the abscissa is azimuthal angle and the ordinate is distance; whereas the B mode display in the ultrasonic apparatus means any mode of display in which the echo intensity is represented by the brightness of cathode ray spots.

The scanning device of the ultrasonic transducer consists of (1), linear scanner; (2), sector scanner; (3), PPI scanner; and (4), compound scanner, consisting of any combination of the other three.[69] In Figure 20 is seen one of the actual systems[70] of its block diagram in the case of a linear sector compound scanning. In a suitable water vessel, the ultrasonic probe; i.e.a transducer, repeats the sector swing which is driven by a motor (1), and at the same time, the probe makes a horizontal displacement driven by the other motor (2). The position and azimuthal direction of the probe are given to the cathode ray tube through a signal generated by a linear potentiometer and a three phase resolver respectively, as is easily seen in the diagram.

In recent years, a scanning mode called contact compound scanning has appeared in medical practice and is useful, especially in the examinations of the head and abdominal organs including the use for obstetrics. The transducer is manually made to swing and displace while it keeps a sliding contact on the skin of the patient.

There are two systems for positioning the ultrasonic probe. One is as shown in Figure 21a. The position of the

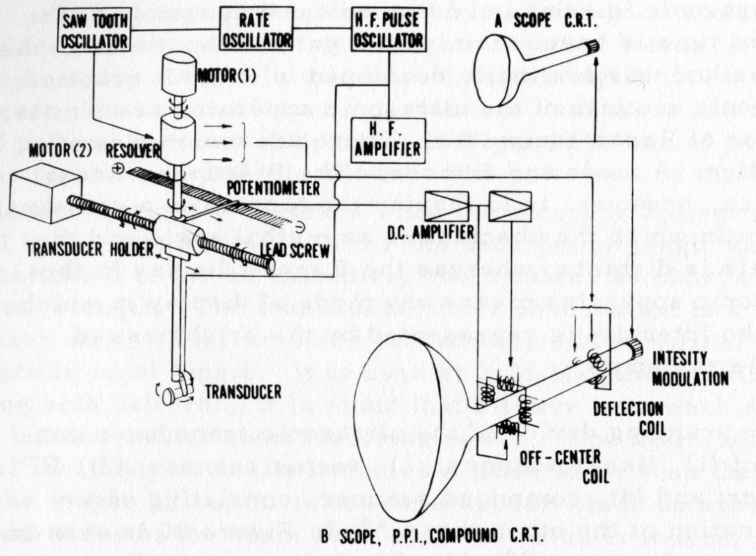

Figure 20. Linear Sector Compound Scanning (Kikuchi, et al.)

© Ebina, T.,* Kikuchi, Y., Oka, S., Tanaka, M., Kosaka, S., Uchida, R., and Hagiwara, Y.: The diagnostic application of ultrasound to the diseases in mediastinal organs--Ultrasono-Tomography for the Heart and Great Vessels(First Report), Science Reports of the Research Institutes Tohoku University Series C (Medicine), Reports of the Res. Inst. for Tuberculosis Leprosy and Cancer, Vol. 12, No. 1, March, 1965, P. 58. P. 60, Figure 3. * Toshiaki Ebina, Prof. Emeritus, Res. Inst. for Tuberculosis and Leprosy, Tohoku University, Sendai, Japan.

ultrasonic probe with reference to a fixed point, 0, is represented by a pair of electric voltages that are produced by a potentiometer device so as to be proportional to x and y, respectively. The direction of the ultrasonic beam is represented by another voltage proportional to the angle, θ. The voltages x and y determine the origin of the time sweep on a cathode ray screen, and the voltage θ gives the direction of the sweep. The other system of positioning is as shown in Figure 21b. The voltages x and y in this case are constructed in such a way that:

$$x = \overline{AB} \sin\alpha + \overline{BC} \sin\beta$$

and

$$y = -\overline{AB} \cos\alpha + \overline{BC} \cos\beta$$

where $\overline{AB}$ and $\overline{BC}$ are the proportional constants representing the lengths of the two mechanical arms which hang the probe from the fixed point, A. The voltages proportional to sin α, sin β, etc., are produced by certain resolvers or special potentiometer devices. The beam direction θ is the same as described above. In Figure 22, an actual system of the mechanical arms and resolvers are shown, for example. At the right tip is a cylindrical ultrasonic probe with which a physician makes manual contact onto the patient's skin. Figure 23 shows the entire equipment with another type of manual compound scanner.

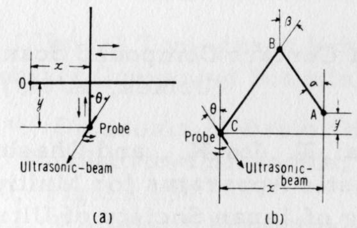

Figure 21. Two Kinds of Positioning Coordinates for Contact Compound Scanning.

© Kikuchi, Y., and Okuyama, D.: Ultrasonic Diagnostics in Medicine, II, Journal of Acoustical Society of Japan, Vol. 27, No. 11, P. 579, 1971, P. 580, Figure 11.

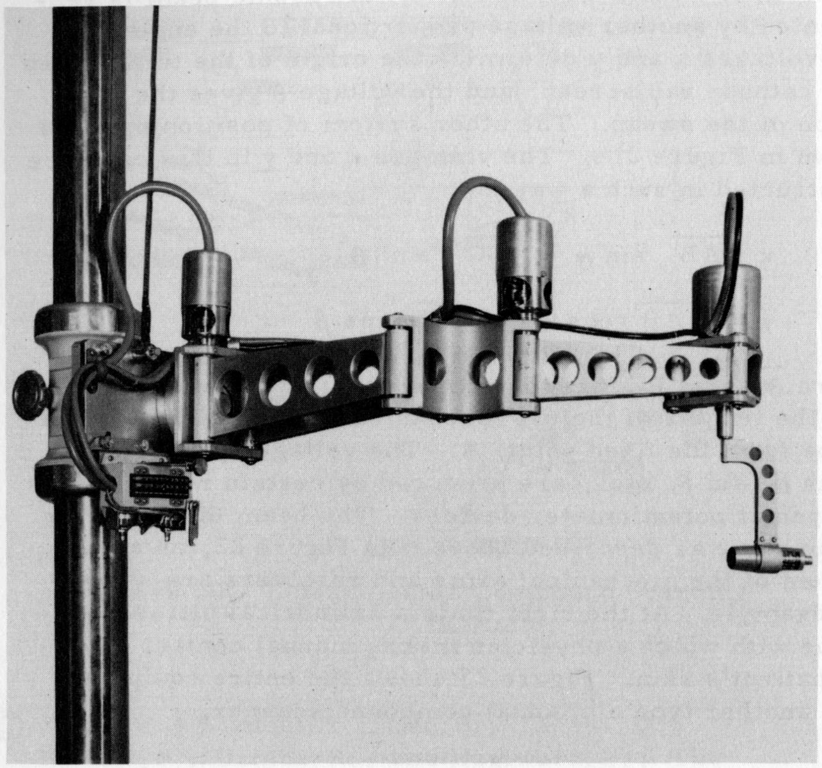

Figure 22. A Contact Compound Scanner (Uchida, et al.)

© Uchida, R., * Wagai, T., Ito, K., and Ohashi, H.: Ultrasonic Diagnostic Apparatus for Multiple Usage, Proc., 4th Meeting of Japan Society of Ultrasonics in Medicine, P. 31, Nov. 1963, P. 32, Figure 5.

* Rokuro Uchida, Japan Radiation and Medical Electronics, Inc., 6-22-1, Murei, Mitaka, Tokyo, Japan.

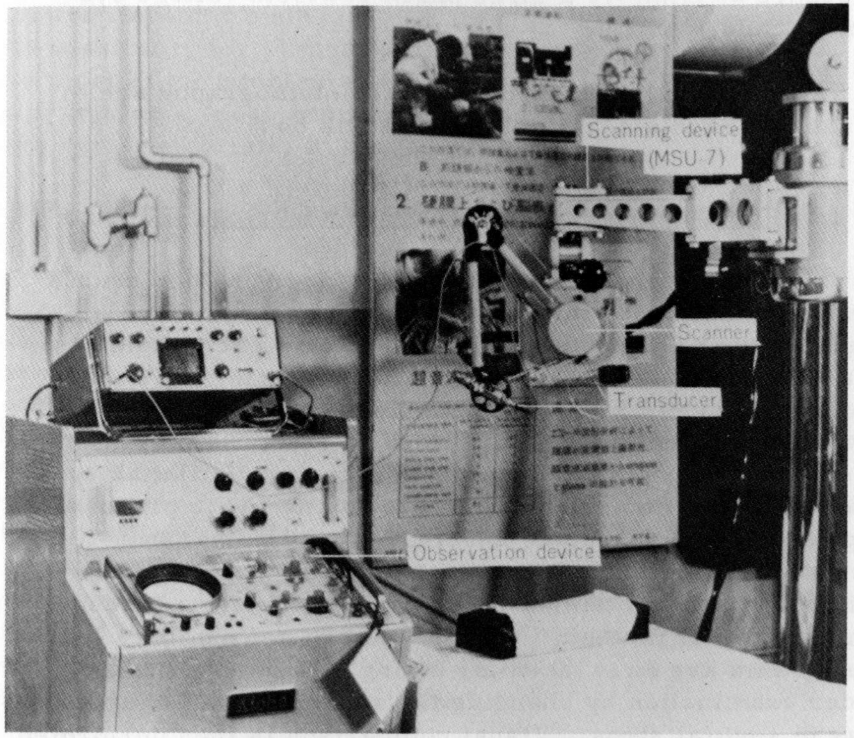

Figure 23. A Clinical Tomograph Setup of a Contact Compound Scanning Type. (Tanaka)

© Tanaka, K. *: Diagnosis of Brain Disease by Ultrasound, Shindan-to-Chiryo Sha, Ltd, Tokyo, 1969, P. 120, Figure 134.

* Kenji Tanaka, Professor Emeritus, Juntendo U. School of Medicine, 2-1-1, Hongo, Bunkyo-Ku, Tokyo.

Actual circuits which treat the azimuthal angle θ are not simple in any scanning system because θ must control the cathod ray sweep in combination with the lapse of time after each pulse transmission. The system already shown in Figure 20 employs a three phase sawtooth voltage system for this function.

Systems for the synchronized cardiotomography are to be described in a separate paper in this book.

2. For Quantitative Examination in Clinics

As has been described in this paper, no one doubts nowadays that ultrasonic examination has become one of the important diagnostic methods in nearly every field of clinical medicine. Insofar as qualitative examination is concerned, it can be said that the method has been well established. As to quantitative description of the examination, however, many research workers are still making efforts toward that establishment. As one of the proposals applicable to tomographic description, the author has introduced a method of quantitative comparison of patterns, and has tentatively named the method "sensitivity graded tomogram pair method."[71] A set of several numbers of tomograms are first taken for a single part of an organ under examination by changing the sensitivity of the apparatus in several steps. Visual comparison is then performed by physicians with any pair among the several tomograms. Targets with higher reflection can appear in both tomograms of the pair, but those with lower reflection cannot; thus, the pair contains much important information for diagnostic judgment. Another pair obtained at another pair of sensitivity settings will do the same. It has been recommended that a set of four tomograms be taken for a single pathological part in six dB steps which may be considered optimum in most of the current diagnostic apparatus. At present, this method is spreading widely among various clinics. Examples of the successful results are shown by Dr Wagai in a separate paper.

V. SOME NEW TECHNIQUES PROPOSED FOR SOLVING PROBLEMS IN ULTRASONOTOMOGRAPHY

There are many requirements and problems that arise from clinics for the improvement of ultrasonotomography toward its wider applicability, its compact instrumentation, its possibility for clinical usage with reduced time, etc. As most of the problems relate to transducers and/or display systems, new technical proposals concerning them are reviewed briefly in the following:

1. High Speed Scanning of Ultrasonic Beams:

Methods for high speed scanning of ultrasonic beams have been considered in accordance with some clinical needs. Asberg[72] has been proposing a high speed sector scanning of a focusing mirror system receiver for obtaining an ultrasonic cinematogram of the living heart. Pätzold and others[73] have developed another high speed scanning system such as is shown in Figure 24. A focusing transducer is made to make a rapid circular motion around the focal line of a paraboloid reflector. The whole system is covered by a plastic sheet and filled with a certain liquid medium. The surface of the sheet is attached to the patient's skin. The reflected focused beam then linearly scans the subject. It is reported that one tomogram of a subject's section of 14 cm x 16 cm in size can be displayed within 60 ms.

As an alternative, Uchida and others[74] have recently proposed a completely stationary system. The idea is as follows: 200 strip transducers are arranged on a plane to form an arrayed probe as shown in Figure 25; and any 20 neighboring transducers are chosen to make an ultrasonic beam. By switching this choice successively, the beam linearly scans the subject on whose skin the arrayed probe is attached, in a stationary manner. An ultrasonotomogram of a sponge-test piece is shown in Figure 26 with its illustration. For a cross-section of 16 cm x 25 cm, it takes about 60 ms to make a tomogram; for a smaller section, however, the required time can be easily shortened.

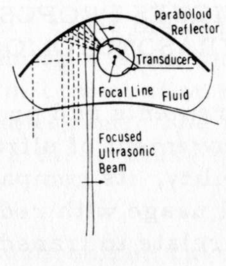

Figure 24. A High Speed Scanner by Means of a Paraboloid Mirror System

Ⓒ Pätzold, J., * Kranse, W., Kresse, H. and Soldner, R.: Present State of an Ultrasonic Cross-section Procedure with Rapid Image Rate, IEEE Trans. on Bio-Medical Engineering, P. 263, 1970, P. 263, Figure 1.

* J. Pätzold, Siemens Aktiengesellschaft, Bereich Medizinesche Technik, Erlangen, Germany.

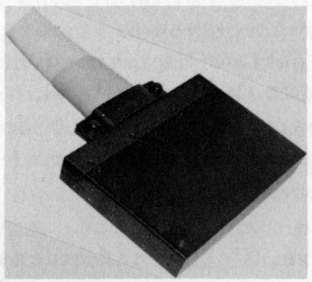

Figure 25. A High Speed Scanner of a Stationary Transducer System. (Uchida, et al.)

Ⓒ Uchida, R., * Hagiwara, Y. and Irie, K.: Electronic Scanning of Ultrasonic beam for Medical Diagnostic Apparatus (a tentative English translation from the Japanese title), Report of Japan Society Ultrasonics in Medicine, 19th Meeting, No. 19-33, 1971 (in Japanese), P. 65, Figure 2.

* R. Uchida, Japan Radiation and Medical Electronics, Inc., 6-22-1, Murei, Mitaka, Tokyo, Japan.

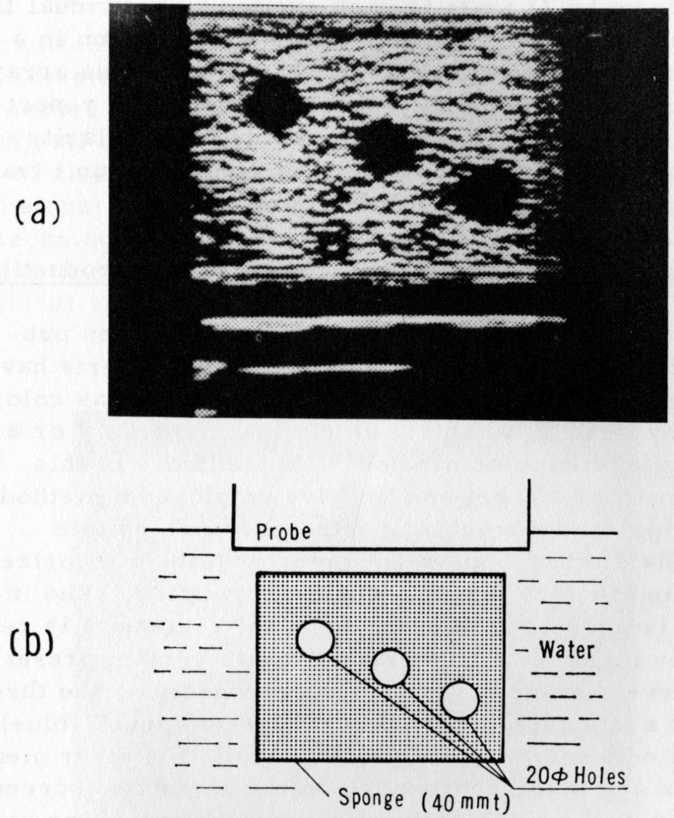

Figure 26. Test Tomogram Obtained by High Speed Scanning. (Uchida)

Ⓒ Uchida, R., Hagiwara, Y., and Irie, K.: Electronic Scanning of Ultrasonic Beams for Medical Diagnostic Apparatus (a tentative English translation from the Japanese title), Report of Japan Society Ultrasonics in Medicine, 19th meeting, No. 19-33, 1971, P.66(a)-Figure 10, (b), Figure 8.

Methods of electronic scanning have also been considered for use in medical ultrasonics. Somer[75] reported a prototype experiment in which every unit transducer in an array made up by 21 units is each excited by individual local oscillators which are to make pulsed oscillation in a proper time sequence so the ultrasonic beam of the array can make a sector scanning. Okujima[76] recently reported an idea of a high speed scanning technique in a polarity correlation principle in which a fewer number of unit transducers are required for sublobe suppression.

2. Tomographic Display, Recording and Reproduction

Recently, several new techniques have been published on tomographic display. On one hand, efforts have been made toward color display by such a means as color-photography itself,[77] a certain electronic method,[78] or a certain digital computer method[79] The authors of this computer method, Yokoi and Ito have employed a method of converting analog signals of ultrasonic echoes into digital signals through an A-D converter to be memorized by a certain memory device. On memorization, echo information is processed so that every echo intensity is represented by three bit signals according to certain preset intensity levels of eight steps. On reproduction, the three-bit signals are applied on R (red), G (green) and B (blue) electrodes of a color kinescope tube, while the other memorized data are being reproduced on the kinescope screen. An example of the black and white reproduction of the color tomogram is shown in Figure 27. The authors have claimed that a single tomogram thus obtained in color presentation has information on intensity level simultaneously and that the sensitivity-graded pair method will be therefore needless in the clinic.

On the other hand, Baum[80] recently disclosed a quantized ultrasonography in which the usual tomogram of extended dynamic range is processed by a certain image quantizer that prints out each isodensitometric area in a code form. Then an acoustic contour map can be prepared

for differential diagnosis. In Figure 28 is shown an example of the quantized image of the human orbit, clearly showing tumor tissues by black masses at Arrows A and B.

Robinson[81] at MIT, recently published a three-dimensional display of ultrasonic data for medical diagnosis. The method is based on a picture digitization and the use of a digital computer. The paper demonstrates the formation of a section at any arbitrary angle through the three-dimensional information. The author has indicated that this technique is to be incorporated in acoustical holography or other means than the current pulse-echo methods used in clinical routine.

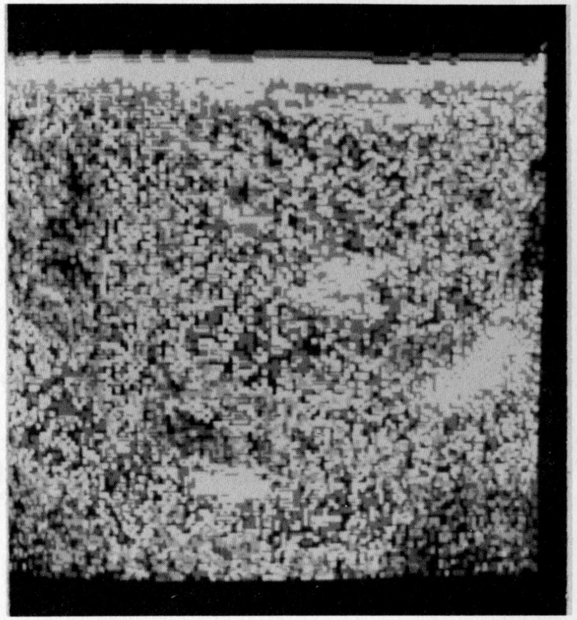

Figure 27. A Tomogram of the Liver and Gall Bladder (Yokoi) (The original picture was in color)[79]

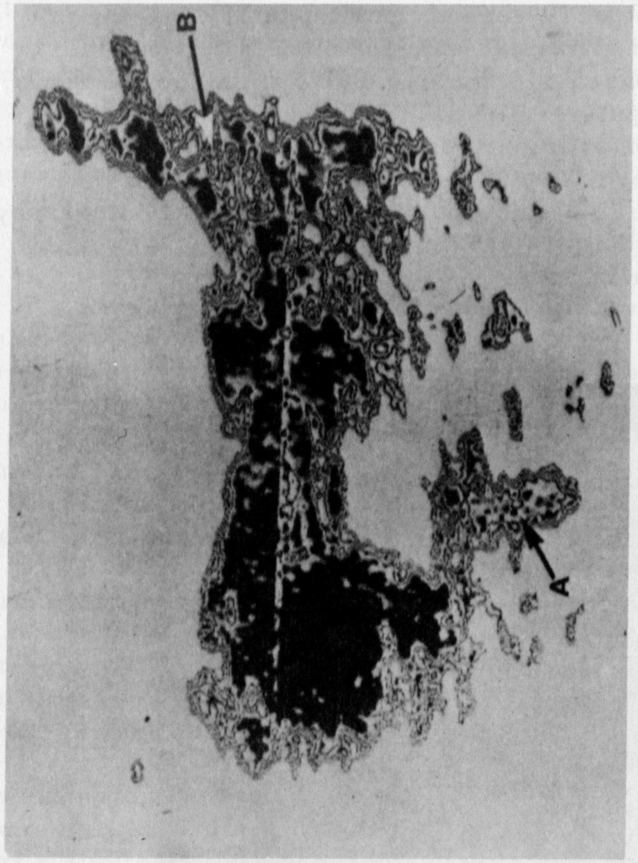

Figure 28. A Quantized Image of the Human Orbit (Baum)

© Baum, G.: Quantized ultrasonography, ultrasonics, P.14, January, 1972. P. 15, Figure 2.

As to video recording techniques for ultrasonotomography, Ide and others [82] have been developing a system for clinical use. Kikuchi, Okuyama and others [83] are engaged in developing a system of multiple information recording and reproduction which is applicable to the synchronized tomography for the living heart. Details of the above two systems will be described in separate papers.

VI. SUMMARY

In this paper, the historical background of ultrasonic diagnostics was first reviewed briefly with some descriptions of A-mode display used in the early stage of the echo-method development. In the next place, the development of ultrasonic tomography was outlined. In Chapter II, the acoustical proportions of biological tissues were briefly summarized in such a way that anyone can reach the outlines of the concept easily. In Chapter III, recent findings concerning the ultrasonic pulse technique were described. In Chapter IV, present systems for ultrasonotomography were explained, together with the concept of sensitivity-graded tomograms, one of the useful kinds of software in medical diagnostics. In the last chapter, some new techniques were shown that are relevant to ultrasonic tomography. These techniques are being developed in this field toward the advancement of ultrasonic diagnostics.

The author hopes that this paper give some knowledge concerning the technique of ultrasonic imaging of human organs, currently used in clinics, so the knowledge would be of use in future development of holographic techniques that will be applicable to clinical medicine.

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NEW FORMS OF ULTRASONIC AND RADAR IMAGING

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ABSTRACT

Recent developments in synthetic aperture techniques to be described, which are applicable to both ultrasonic and radar imaging, include: a solution to the ambiguity problem, creating end-fire gain and synthetic gain against moving targets for both monostatic and bistatic systems. Other subjects to be discussed include acoustic kinoforms, real-time holographic detection of concealed weapons, synthetic aperture hologram interferometry, and new developments in (1) real-time imaging with sonar arrays comprising thousands of elements, and (2) seismic imaging.

EARLY ACOUSTIC HOLOGRAMS

If one defines an acoustic hologram as the photographically recorded interference pattern between a set of coherent sound waves of interest and a coherent reference wave generated by the same source, then one can say that the first acoustic holograms were made at the Bell Telephone Laboratories in 1950.[1] In that work, only the holograms (the interference patterns) were of interest and no wave reconstruction was performed; nevertheless, the method employed in recording the interference patterns is still useful today.

The interference patterns were made visible by photographic scanning, using a scanning microphone. To generate